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Assessment of the ankle muscle co-contraction during normal gait: A surface electromyography study

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ABSTRACT

The study was designed to assess the co-contractions of tibialis anterior (TA) and gastrocnemius lateralis (GL) in healthy young adults during gait at self-selected speed and cadence, in terms of variability of onset–offset muscular activation and occurrence frequency. Statistical gait analysis (SGA), a recent methodology performing a statistical characterization of gait by averaging spatio-temporal and EMG-based parameters over numerous strides, was performed in twenty-four healthy young adults. Co-contractions were assessed as the period of overlap between activation intervals of TA and GL. Results showed that GL and TA act as pure agonist/antagonists for ankle plantar/dorsiflexion (no co-contractions) in only 21.3 ± 8.2% of strides. In the remaining strides, statistically significant ($p < 0.05$) co-contractions appear in early stance (29.2 ± 1.7%), mid-stance (32.1 ± 18.3%) and swing (62.2 ± 2.0%). This significantly increased complexity in muscle recruitment strategy beyond the activation as pure ankle plantar/dorsiflexors, suggests that co-contractions are likely functional to further physiological tasks as foot inversion, balance improvement, control of ankle stability and knee flexion. This study represents the first attempt for the development in healthy young adults of a "normality" reference frame for GL/TA co-contractions, able to include the physiological variability of the phenomenon and eliminate the confounding effect of age.

1. Introduction

Muscle co-contraction is the simultaneous contraction of agonist and antagonist muscles crossing a joint (Olney, 1985). Co-contraction generally-held purpose is to augment ligament function in maintenance of joint stability, providing resistance to rotation at a joint and equalizing pressure distribution at joint surfaces (Baratta et al., 1988). Muscle co-contraction has an important role in movement regulation during motor learning activities, enhancing joint stability (Darainy and Ostry, 2008). Thus, muscle co-contraction should be a crucial factor to consider during motor rehabilitation (Den Otter et al., 2006).

Different gait studies reported the presence of co-contraction of ankle plantar-flexor and dorsi-flexor muscles in healthy subjects; although the main task of these muscles is to oppose each other in action causing sagittal plane movement, co-contractions throughout the stance phase were observed (Frost et al., 1997; Olney, 1985; Peterson and Martin, 2010). It was shown that co-contraction varies during the gait cycle (GC), being higher during weight acceptance (Falconer and Winter, 1985) and stance-to-swing transition, and lower at mid-stance (Olney, 1985). A recent analysis including tibialis anterior (TA) and gastrocnemius lateralis (GL), showed that amplitude and timing of muscle co-contraction are correlated with age and velocity in healthy adult walking (Hortobagyi et al., 2009). This matches with the observation that older adults exhibit greater activation of TA and soleus during mid-stance at different walking speeds, suggesting increased co-contraction across the ankle (Schmitz et al., 2009). Franz and Kram (2013) confirmed the greater co-contraction of gastrocnemius and TA and of soleus and TA in older vs. young adults during level walking, and added that neither uphill nor downhill walking affected it. Thus, ankle-muscle activations and co-contractions change with age, suggesting the suitability of developing reference electromyographic databases, separated for young and old subjects.

The complex relationship among the activations of agonist/antagonist muscles that underlies muscle co-contraction is generally explored by surface electromyography (sEMG) (da Fonseca et al., 2004). Since a stride-to-stride variability in sEMG profiles from GL and TA was reported (Di Nardo et al., 2013; Winter and Yack, 1987), a variability in ankle-muscle co-contraction is also

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expected. Thus, besides the effect of age, the development of a reliable reference frame of ankle-muscle co-contraction in healthy adults should consider also the natural variability associated with muscle activation during free walking. A tool like this, able to include the natural variability of the phenomenon and to eliminate the confounding effect of age, would be useful for discriminating physiological and pathological conditions. This can be achieved by recording the electromyographic signal over numerous steps per subject and analyzing data by means of a technique which allows to organize a large amount of signals and to summarize the results, as the statistical gait analysis (SGA) (Agostini and Knaflitz, 2012). SGA is a recently developed methodology, which performs a statistical characterization of gait by averaging spatial–temporal and EMG-based parameters over numerous strides, during the same walking trial. The main advantage is that SGA allows to manage a large amount of data from a single subject, to extract a large amount of information on the variability of sEMG signal and to report the summarized results in a user-friendly way. SGA provides, indeed, mean muscle-activation intervals and occurrence frequency, allowing to directly compare SGA results with those reported in studies performed with classic techniques (Sutherland, 2001). The main disadvantage is that SGA requires a very high number of strides to be run and, consequently, the signal acquisition procedure could be more arduous than in classic methodologies (Sutherland, 2001).

Thus, the aim of the present study was the quantitative assessment of TA/GL co-contractions in healthy young adults during gait at self-selected speed and cadence, in order to develop a reference frame in terms of variability of onset–offset muscular activation and occurrence frequency. Muscular co-contraction was assessed as the overlapping period between activation intervals of agonist and antagonist muscles (Den Otter et al., 2006), computed by robust techniques for detection of muscle activation intervals (Bonato et al., 1998; Staude et al., 2001), and specific tools for SGA. To avoid the effects of shod walking and variations in heel height on EMG signals (Murley et al., 2009), the study was performed on subjects walking barefoot.

2. Materials and methods

2.1. Subjects

Twenty-four healthy young adult volunteers (12 females and 12 males) were recruited (age 23.9 ± 1.9 years; height 173 ± 10 cm; weight 62.7 ± 13.0 kg; body mass index (BMI) 20.6 ± 2.2 kg m^{-2}). Exclusion criteria included history of neurological disorders, orthopedic surgery, acute/chronic knee pain or pathology, BMI > 25, or abnormal gait determined observationally by a licensed physical therapist, specialized in gait analysis. Participants signed informed consent.

2.2. Signal acquisition

Signals were acquired (sampling rate:2 kHz; resolution:12 bit) and processed by the multichannel recording system, Step32 (Version PCI-32 ch2.0.1. DV) DemItalia, Italy. Each subject was instrumented with foot-switches, knee electro-goniometers and sEMG probes on both lower limbs. Three foot-switches (Step32, DemItalia, Italy; size:11 × 11 × 0.5 mm; activation force:3 N) were attached beneath heel, first and fifth metatarsal heads of each foot. An electro-goniometer (Step32, DemItalia, Italy; accuracy:0.5°) was attached to the lateral side of each lower limb for measuring knee-joint angles in sagittal plane.

sEMG signals were detected with single-differential sEMG probes with fixed geometry constituted by Ag/Ag-Cl disks

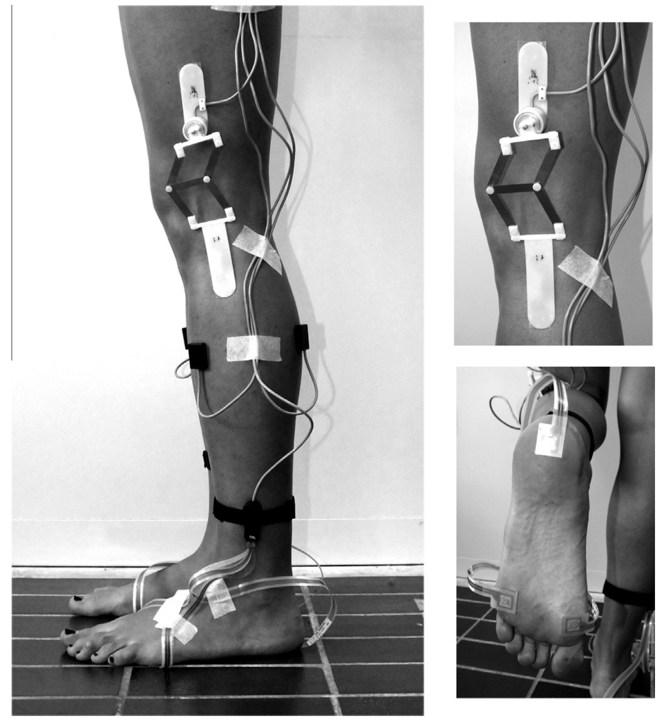

Fig. 1. Participant set-up.

(manufacturer: DemItalia, size:7 × 27 × 19 mm; interelectrode distance:12 mm, gain:1000, high-pass filter:10 Hz, input impedance > 1.5 GΩ, CMRR > 126 dB, input referred noise ⩽ 1 μV_{rms}). EMG signals were further amplified and low-pass filtered (450 Hz) by the recording system; an overall gain, ranging from 1000 to 50,000, could be chosen to suit the need of the specific muscle observed (Agostini et al., 2010).

Before positioning the probes, skin was shaved, cleansed with abrasive paste and wet with a soaked cloth. To assure proper electrode–skin contact, electrodes were dressed with highly-conductive gel. Probes were applied over GL and TA, following the SENIAM recommendations for electrode location-orientation over muscles with respect to tendons, motor points and fiber direction (Freriks et al., 2000). Participant set-up is shown in Fig. 1. Then, subjects were asked to walk barefoot overground for 5 min at natural speed and cadence, following the path described in (Di Nardo and Fioretti, 2013).

2.3. Signal processing

Footswitch signals were debounced, converted to four levels, Heel contact (H), Flat foot contact (F), Push-off (P), Swing (S), and processed to segment and classify the different GCs (Agostini and Knaflitz, 2012).

Goniometric signals were low-pass filtered (FIR filter, 100 taps, cut-off frequency 15 Hz) (Agostini et al., 2010). Knee angles in sagittal plane along with sequences and durations of gait phases derived by basographic signal, were used by a multivariate statistical filter, to detect outlier cycles like those relative to deceleration, reversing, and acceleration. Cycles with improper sequences of gait phases (i.e. different from H–F–P–S sequence), not corresponding to straight walking and with abnormal timing and knee angles, with respect to a mean value computed on each single subject, were discarded (Agostini and Knaflitz, 2012).

sEMG signals were high-pass filtered (FIR filter, 100 taps, cut-off frequency of 20 Hz) and processed by a double-threshold statistical detector, allowing a user-independent assessment of muscle